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Virtual Reality Interaction and Physical Simulation

Interactive physically-based simulation of catheter and guidewire

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Abstract

For over 20 years, interventional methods have improved the outcomes of patients with cardiovascular disease or stroke. However, these procedures require an intricate combination of visual and tactile feedback and extensive training periods. An essential part of this training relates to catheter or guidewire manipulation. In this paper, we propose a composite model to realistically simulate a catheter, a guidewire or a combination of both. Where a physics-based simulation of both devices would be computationally prohibitive and would require to deal with a large number of contacts, we propose to address this problem by replacing both objects by a composite model. This model has a dual visual representation and can dynamically change its material properties to locally describe a combination of both devices. Results show that the composite model exhibits the same characteristics of a catheter/guidewire combination while maintaining real-time interaction.

Keywords: Physically based simulation; Interactive animation; Catheter/guidewire simulation; Composite model

1. Introduction

Over the last 20 years, interventional methods such as angioplasty, stenting, and catheter-based drug delivery have substantially improved the outcomes for patients with cardiovascular or neurovascular disease. However, these techniques require an intricate combination of tactile and visual feedback, and extensive training periods to attain competency. Traditionally, the best training environments on which to learn the anatomic–pathologic and therapeutic techniques have been animals or actual patients. Yet, the development of computer-based simulation can provide an excellent alternative to traditional training. To reach this goal, most aspects of the real procedure need to be simulated, specifically catheter[1] and guidewire[2] manipulation. In a real procedure, after puncturing the femoral artery, a guidewire–catheter combination is advanced under fluoroscopic guidance (sequence of X-ray images) through the iliac arteries and aorta then into the aortic arch just above the heart. Then the catheter or guidewire is advanced into the neurovascular network or the coronary arteries. Once in place, various therapeutic devices can be manipulated to the target location using fluoroscopic guidance.

A few computer-based training systems focusing on interventional radiology have been developed or commercialized as of today [1–3]. These systems include interactive models of catheters but do not deal with the complex interactions between a device and the vessels or between catheter and guidewire. We propose a composite model of catheter/guidewire that handles these two problems. A physics-based device model is used to create a realistic and accurate behavior, while a specific visualization technique is also proposed to render both objects from one unified model. Also, an update of the material properties and other intrinsic parameters of the model is performed dynamically to describe a combination of both devices using a unique model.

This paper is organized as follows: previous work is discussed in Section 2, then our proposal is outlined

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[1] A hollow, flexible tube inserted into a vessel to allow the passage of fluids or other devices.

[2] A flexible wire positioned in a vessel for the purpose of directing the passage of a catheter.

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in Section 3, and preliminary results are presented in Section 4.

2. Previous work

Previous work in the field of interventional radiology simulation has focused on the development of complete systems [3,1], visualization [4] and anatomical modeling [5]. In [1] the authors simulate the catheter using a linear elasticity FEM-based representation. This technique assumes that the catheter moves with small displacements to remain valid with linear elasticity theory. In [3] the catheter simulation is based on a multi-body system composed of a set of rigid bodies and joints. This model can be seen as a macroscopic approximation of a continuous model at some level. Twist motion on the extremity is transmitted to all rigid bodies via the links and bending is represented by angular springs placed between consecutive rigid links. This discrete model permits a good approximation of a catheter but requires many small links to represent a high degree of flexibility, thus leading to increased computation cost.

Other work outside of the area of interventional radiology has focused on one-dimensional deformable models. For instance, Lenoir et al. [6] proposed to simulate a surgical thread in real-time using a dynamic spline. This continuous physics-based model is based on Lagrange equations combined with spline geometry. While this model does not use continuous energies, Nocent et al. [7] have proposed a specific continuous energy to handle the stretching deformation of a dynamic spline. Although this work is a first step toward the simulation of a catheter, it lacks bending and twisting energies to be able to represent accurately such a device. Another way to model one-dimensional deformable objects is to use the Cosserat theory, as proposed by Pai [8]. This model is static and takes into account all possible deformations of a one-dimensional object, however contact handling with such a model is difficult. For catheter navigation, where collisions occur continuously along the length of the device, this is a critical issue.

Finally, some recent work directly related to the simulation of a catheter or guidewire has been proposed by Cotin et al. [9]. This physics-based model consists of a set of connected beam elements that can model bending, twist and other deformations in real-time. The simulation is based on a static finite element representation. The approach described in the following sections is based on this work. Different aspects of this model are presented in Section 3.1 while our contribution begins in Section 3.1.1.

3. Composite catheter–guidewire model

The main idea of our method is to decompose the animation of the catheter/guidewire combination into a physics-based component and an animation component. By simulating only one model, we avoid the problems of handling the numerous contact between the catheter and the guidewire, since both devices are co-axial. This section is organized in four subsections describing respectively: the physics-based simulation of the composite model, the animation part, the rendering of the two different objects and the physical interaction between both objects.

3.1. Physics-based component

As previously mentioned, our approach is based on a physics-based model (described in details in [9]). The model is defined as a set of beam elements. Each element has six degrees of freedom, three degrees of freedom in translation and three in rotation. The model is continuous and the equations are solved using a finite element approach. Since each element can bend, a lower number of elements is required to represent the catheter than with a rigid body approach [3]. The choice of a static over a dynamic model was made since the catheter or guidewire navigate inside blood vessels where the blood induces a damping factor. Under this assumption, the equations of the mechanical system can be written as

$$[K]U = F, \tag{1}$$

where U represents the degrees of freedom, F the forces and torques applied on each nodes and $[K]$ the stiffness matrix of the system.

Because of the significant flexibility of a catheter or guidewire with the high curvature of the vascular system, the model generates geometric non-linearities which cannot be handled by a linear model. This can be overcome by defining $[K]$ as a function of the degrees of freedom U:

$$[K(U)]U = F.$$

The system is then solved using an incremental approach [9] which permits faster computation times compared to other techniques. Additionally, the system of equations can be rewritten by decomposing the model as a set of substructures [9].

To control the deformation and navigation of the device, we use a combination of external forces and boundary conditions. Since the device (catheter or guidewire) is manipulated by the interventional radiologist using only a combination of translations and rotations about the main axis of the device, this is taken into account in the model via a set of boundary conditions (BC). Those constraints are transmitted along the entire model during its deformations, according to the material properties.

As the device is inserted within a vessel, we need to deal with collisions with the vessel's walls. This contact problem can be split into the detection part and the response part. The next two subsections describe our methodology for handling these two problems.

3.1.1. Collisions detection

We suppose that we have access to an highly detailed vascular model. Such a model can be extracted from a CT scan of a patient and reconstructed in three dimensions using a set of image processing and geometry tools. As an example, the reader can refer to [5]. Considering the three-dimensional dataset, we decompose the vascular model as a set of connected segments described as a graph (as illustrated in Fig. 1). Each node of the graph is linked to a set of triangles describing locally the surface of the vascular model. Combining this with a search algorithm based on temporal coherence, this reduces the search space of triangles for the collision detection process.

An example is given in Fig. 1(a), where a node of the catheter initially inside segment *s4* moves outside this segment, requiring the collision detection algorithm to update its segment number. This method starts by checking if the node is still in the same segment, then searches in the nearest neighbor's using the graph of connectivity presented in Fig. 1(b). The function returns the corresponding segment number, in this example: *s6*. If the node does not belong to the set of nearest neighbors, the algorithm searches recursively into the next neighbors with a stop criteria based on a depth limit.

It can be noted that a node can exit the three-dimensional model of the arterial network (according to the collision response detailed in Section 3.1.2). This special case is detected when the recursive search does not find any solution. In that case, the algorithm assigns the segment that keeps the more consistency accordingly to its last segment and its motion. This way, a node cannot jump to a nearly vessel.

The three-dimensional model of the arteries can be multi-resolution and thus can provide different resolution of the mesh. By this way, the collision detection can be done in a selected resolution depending on the remaining time or another criteria (distance of the camera…).

3.1.2. Collisions response

The collision response acts like a pipeline process, by taking the collision detection output, resolving the static system including the effective collisions and giving the new state in output, like schematized in Fig. 2.

The solution is computed iteratively by enabling some collision constraints while disabling others according to a specific algorithm. For each iteration, the method solves the constrained system with a dedicated Gauss–Seidel method (but others techniques can also be employed, like Lagrange multipliers…). We will focus on the enable/disable constraint algorithm instead of the Gauss–Seidel method as it is a classical method of contact resolution (the reader can refer to [10,9] as examples). Usually, it exists a maximum number of iteration, controlling the resolution time. This iterative algorithm converges towards the solution, but if it is stopped prematurely it will return an approximation to the exact solution.

The resolution of this iterative algorithm is based upon linear constraints which correspond to infinite planes of the triangles. If a set of triangles form a convex polyhedra, then all of their contraints are complementary and can be merged together within the system. If on the other hand, the triangles form a concave polyhedra, some triangle planes will restrict the sub-space of available motion for that particular catheter node as shown in Fig. 3(b).

To avoid this problem each triangle of each segment is grouped either in a convex or concave set, as shown in Fig. 3.

All the plane's constraints defined by the convex set are added to the system because they do not restrict the mobility space. We define the term *closest concave triangle* (of a node) as the triangle of the concave set that has the smallest orthogonal projective distance within the interior of the triangle. The closest concave triangle is also added (if one is found) to the constraints set. After a collision response iteration, the algorithm checks if the closest concave triangle of the node is still the same.

- If it is the case, this means that the node has moved on the triangle's sub-space (defined by the orthogonal projection of its area) but did not get farther, so that the computed solution is a valid solution.
- Otherwise, the node has moved outside this volume so that it may now violate another constraint defined by another concave triangle. The algorithm will then iterates by picking another concave triangle.

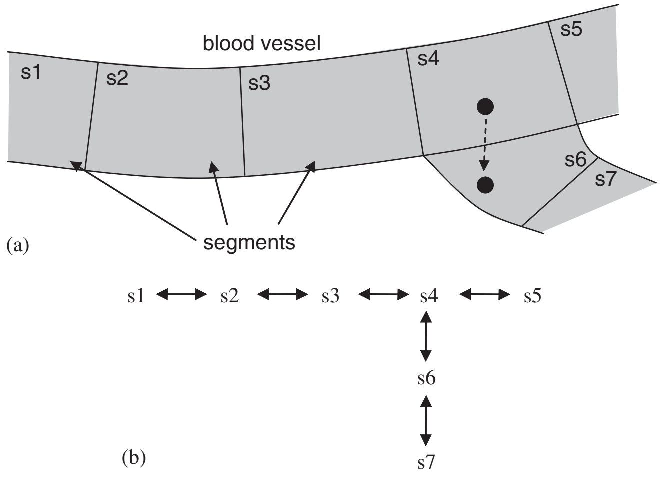

Fig. 1. (a) Decomposition of a vessel into segments. (b) Connectivity graph of the segments.

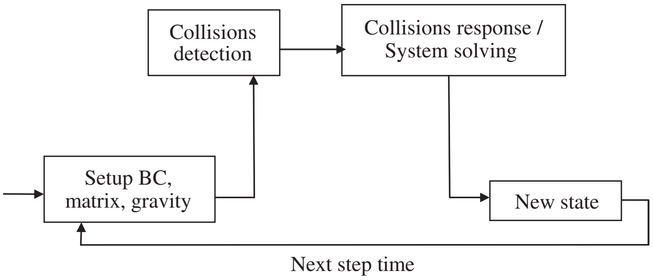

Fig. 2. Automata of a simulation step.

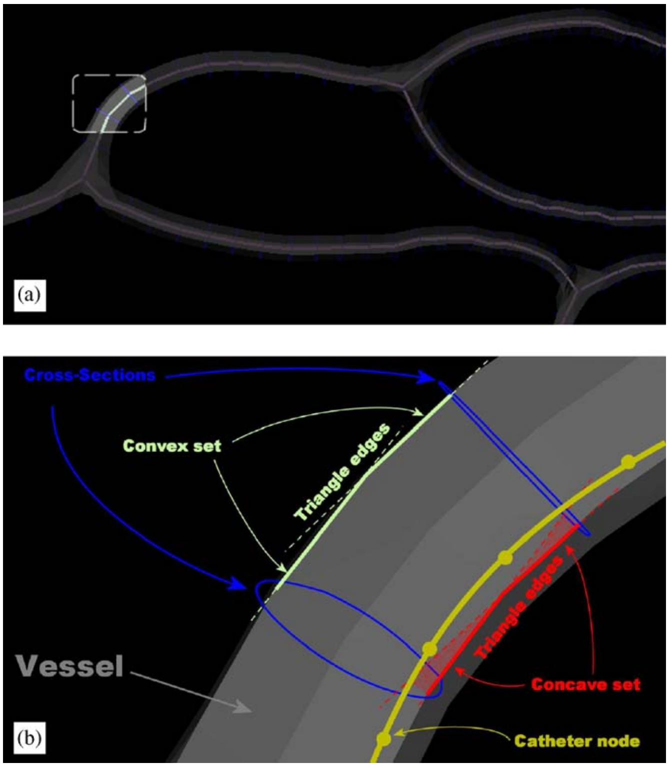

Fig. 3. The triangulation of the vascular model can be classified in two categories: triangles that belong to a convex set and triangles that belong to a concave set. Non-convex sets over-constrain the problem by reducing the solution space. These two cases are handled differently in the collision response algorithm. (a) View of the vascular network and selection of an area of interest. (b) Zoom on the area of interest showing a segment defined by two cross-sections.

The algorithm is schematized in Fig. 4.

Using this approach, nodes can slide within a neighborhood of triangles in only one simulation step, allowing large displacements.

The physical simulation described above is the core aspect of representing a device with our system. However, to animate co-axial models, an animation technique can be used to replicate guidewire/catheter interactions.

3.2. Animation component

A catheter is a hollow device in which enclose a guidewire. In real situations, a movement of the guidewire inside the catheter modifies the behavior of the catheter. This part is linked to the interaction between both devices and will be discussed in Section 3.3. This section only deals with the visual part of the animation technique and is decomposed in two subsections, an animation algorithm and a visualization proposition.

3.2.1. Animation algorithm

The animation part is based on the visual appearance of the catheter and the guidewire, which is needed to distinguished between them. This is the role of a single floating value called *limit*. *limit* determines the position of the guidewire tip relatively to the catheter tip in sense of a curvilinear abscissa, like schematized in Fig. 5.

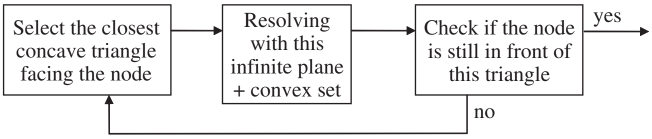

Fig. 4. Automata of the collision response algorithm.

Typically, change of *limit* happens only when the interventional radiologist push or pull the catheter or the guidewire. By doing so, it changes the existing relation between the two devices. The algorithm used to update the *limit* value, supposing that the motion is described by a signed *translation* value assigned to either the catheter or guidewire extremity is:

- *The guidewire is inside the catheter* ($limit \leqslant 0$):

  A motion of the guidewire imposes an update of the *limit* value and checks for an potential effective translation:

```
$limit = limit + translation$
if $limit > 0$ then
    $translation = limit$
else
    $translation = 0$
end if
Apply $translation$ to the physical model extremity
```

  A catheter movement involves the following algorithm:

```
$limit = limit - translation$
if $limit > 0$ then
    $translation = translation - limit$
end if
Apply $translation$ to the physical model extremity
```

- *The catheter moves along the guidewire* ($limit > 0$): reciprocal algorithm.

This *limit* value is useful in the rendering algorithm that is described in the next subsection but also in the interactions between the catheter and the guidewire described in Section 3.3.

3.2.2. Visualization

Both device models are rendered as generalized cylinder [11,12] (cf. Fig. 6). This technique permits to define a smooth surface (Fig. 6(b)) based on a skeleton and a cross-section (Fig. 6(a)). As a result, the smoothness of the real device are kept in the animation. Moreover, this technique permits mapping textures on the cylinder's surface and shows the effective twisting of the device. Furthermore, new technique [13] proposes to optimize the rendering speed of generalized cylinders utilizing graphics hardware.

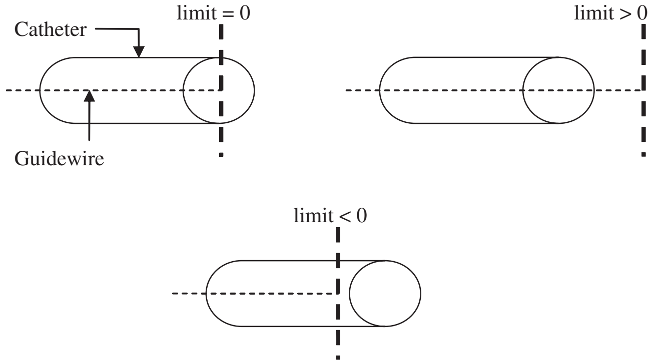

Fig. 5. The three possible configuration for limit value.

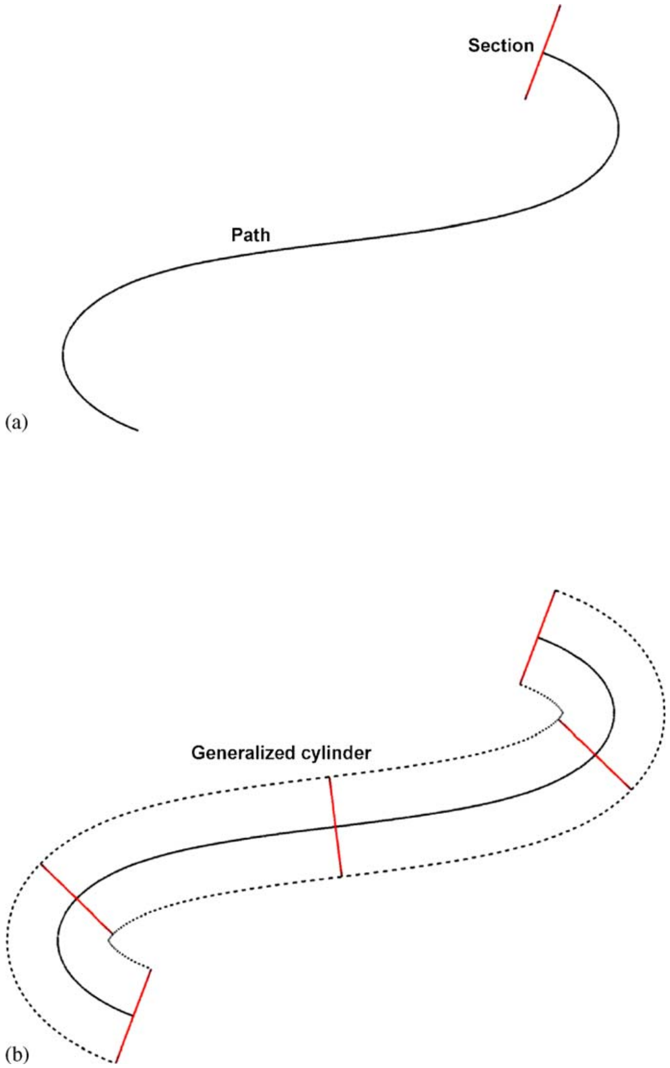

Fig. 6. Generalized cylinder construction. (a) Basic elements of a generalized cylinder. (b) Construction of a generalized cylinder.

This technique is appropriate since the catheter and the guidewire have some important curvature avoiding multiple loop for example, so that the case of self collision does not happened naturally.

The catheter is rendered with a chosen alpha channel to show the progression of a potential device evolving inside. On the other hand, the guidewire is rendered in plain mode because of its solid structure. By this way, both devices can be seen at the same time even if the guidewire is completely inside the catheter.

Generalized cylinder are constructed thanks to a constant circle section and a path defined by the nodes position. The last point of the path is interpolated using the *limit* value.

To exhibit realistic behavior for both objects, a special technique has to be developed for the mechanical interaction between a catheter and a guidewire.

3.3. Catheter/guidewire interaction

When a guidewire is inserted through a catheter or a catheter moves along a guidewire, the overall shape of both the catheter and guidewire is modified due to a change in the bending stiffness and bending moment in the overlapped region. This co-axial region offers a stiffer resistance to transverse loading. We will simulate this meaningful visual cue as a fiber *reinforced* composite material. The transversal stiffness of the overlapped region will be modeled with the well-established empirical expression, the Halpin–Tsai equations [14]:

$$E_{trans} = \frac{E_{cath}(1+\xi\eta f)}{1-\eta f}, \quad \eta = \frac{E_{guide} - E_{cath}}{E_{guide} + \xi E_{cath}}, \tag{2}$$

where f is the ratio, in the overlapped section, of the guidewire volume over the volume of the guidewire–catheter combination; ξ is a function of the material properties and geometry of the instruments. Lookup tables describing typical values of ξ under different composition configurations have been published in the literature [14]. The stiffness for the overlapped section will be updated in real-time and the composite physical model will reflect this change, accordingly. By using this approach, we can represent the catheter/guidewire combination as a particular implementation of our initial physical model. This allows us to avoid computing the collision between two separate objects, as they are treated as a single composite model.

This approach allows the representation of co-axial objects as the modulation of material properties over one base model without additional computation cost.

4. Results

The first result (related to Fig. 7) shows the effectiveness of the physical model. In this simulation, the mechanical model traverses through a vessel network and is simulated with 100 nodes in interactive time (about 45 iterations per second). The pictures exhibit a realistic behavior of the catheter model, especially after a bifurcation of the network. In that case, the model presents characteristic bending just before the bifurcation.

The second result points out the collision detection, that handles the interactions with the vessels. Fig. 8 presents the collision detection at one step of the simulation. The yellow triangles represent the colliding triangles of the vessels. This decomposition into segments enhances the collision detection efficiency. The algorithm is based on local information, so that even if the vessel network was extended, the collision detection would be still efficient.

The last result (presented in Fig. 9) shows the strong interaction between the two nested devices. In this animation, a straight guidewire is inserted into a curved catheter and deformes it. In the top picture, the guidewire is located just before the tip of the catheter so that we can see the natural bending of the tip of the catheter. In the middle picture, the guidewire progresses through the catheter deforming it. In the last bottom picture, the guidewire completely deformed the catheter by overcoming its bending property.

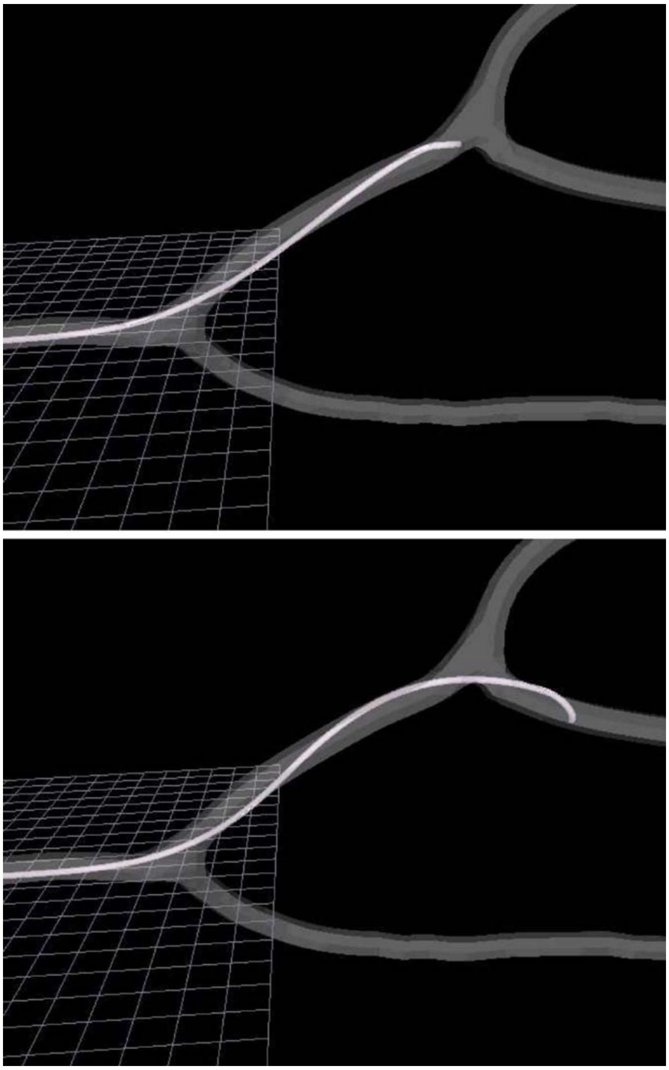

Fig. 7. Two step of a simulation in the vessel network.

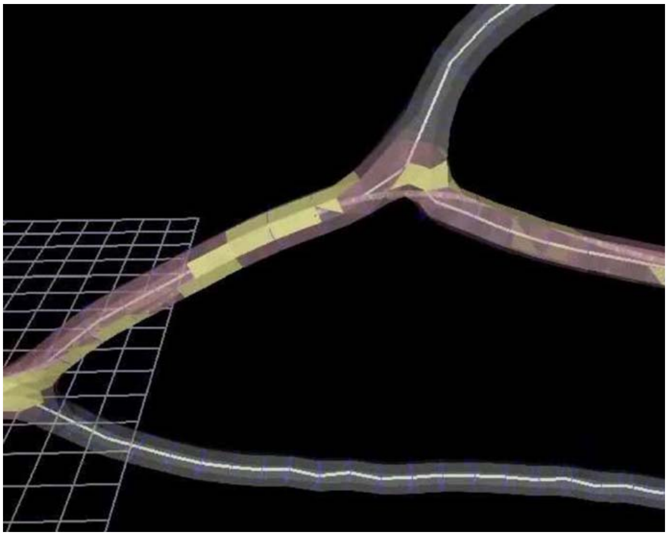

Fig. 8. Collision detection of the model with the vessels.

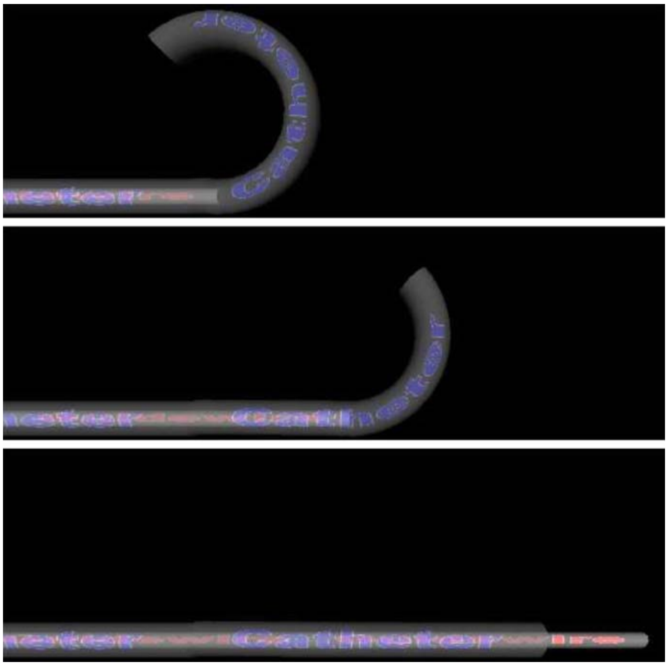

Fig. 9. Guidewire deforming a catheter.

5. Conclusion and future work

This paper proposes an unique model to interactively simulate a catheter and a guidewire. This model exhibits realistic behaviors for the two simulated objects and enhances the model with a specific visualization technique.

Also a composite technique is also presented so that nested device model can be mechanically model efficiently by modulating material property of our core FEM model.

This work can be improved by exploiting the non-linearity of the elastic law by considering the non-linear system and using a minimization technique to resolve the non-linear system. This would provide exact non-linear elasticity instead of linearized one, but it also has to be done in an interactive time, which is an issue. Another improvement concerns the collision process, in this paper, constraints are linear bringing infinite planes, but it can be useful to use quadratic constraint to approximate more accurately the surface. By doing so, a quadratic constraint would cover more space than a linear one, which would result in less constraints.

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